Proceedings of ICMMB-11:
International Conference on Mechanics in Medicine and Biology
April 2-5, 2000 / Maui, Hawaii

MEMS PACKAGING FOR BIOMEDICAL APPLICATIONS

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ABSTRACT

MEMS packaging for biomedical applications can be achieved in several ways. One is complete device encapsulation. Several examples are discussed, including the encapsulation of a MEMS heart cell force transducer in which the encapsulant is composed of four different materials. The second option is a reconfigurable packaging scheme that consists of an inert protective polymer layer that can be moved or broken on command. This type of package facilitates direct sensor access to the environment and preliminary results are presented.

INTRODUCTION

Micro-electromechanical systems (MEMS) technology provides a relatively new, inexpensive way to make sensors. By using silicon-based integrated circuit fabrication techniques chemical, inertial, thermal, and pressure sensors have been miniaturized [1]. MEMS has tremendous growth potential especially in the biomedical sciences. Miniature MEMS biosensors have already been used to measure and interact with a variety of biomolecules, and wireless communication to and from surgically implanted biomedical MEMS devices has been demonstrated [1] [2]. However, advances in MEMS biosensor packaging are fewer in number. This often neglected but important device aspect is necessary to complete a system. Especially in wet, corrosive environments, such as biological fluid, reliable packaging of active devices is extremely challenging.

Two methods of biosensor packaging can be employed: (1) complete encapsulation of active components or (2) reconfigurable encapsulation that allows the active components controlled access to the environment. Complete encapsulation is a more straightforward approach and has proven effective for both standard and MEMS biomedical devices. However, total encapsulation inherently limits the type of sensing that can be performed. Sensors ultimately need access to the environment and will thus require a more sophisticated, interactive package that simultaneously protects the sensor while allowing the required access when needed. Both approaches will be discussed.

ENCAPSULATION PACKAGING

For many years, medical devices such as pacemakers consisted of electronics encased in biocompatible hard shell titanium housings to protect the device from hostile body fluids. Miniature, in-situ sensor systems operating in harsh liquid environments have also involved complete encapsulation. For example, a swallowable temperature sensor has been developed by HTI Technologies Incorporated (*Figure 1*) [3]. The pill is entirely coated in silcone rubber and epoxy which enables it to withstand the acidity and toxicity of the GI tract.

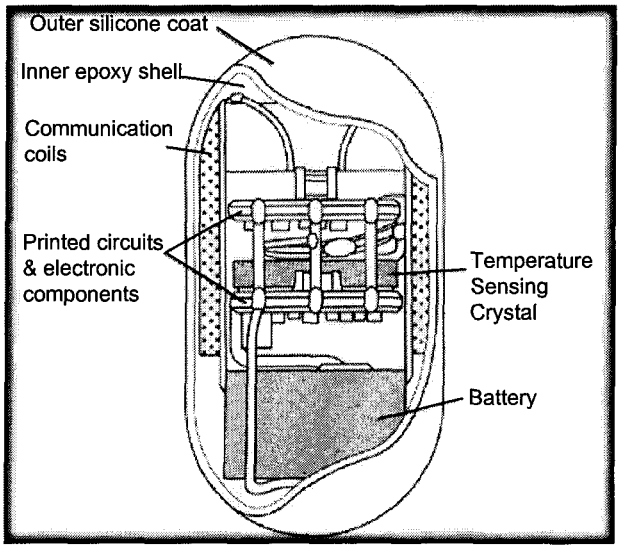

Figure 1. Diagram of a swallowable temperature sensor by HTI Technologies Incorporated [3].

In addition to standard dip coating and spray coating, encapsulation may require custom processing. Ziaie, et. al., used an ultrasonically machined glass capsule that was electrostatically bonded to the substrate to protect the receiver circuitry and hybrid elements of an implantable MEMS nerve microstimulator from body fluids [4].

Encapsulation may also require more than one material. For example, the force generated by an individual heart cell was measured using a completely

encapsulated MEMS device. The packaging required for the chip and supporting electrical interconnects required complete encapsulation using silicon dioxide, epoxy, silicone rubber sealant, and enamel. The force transducer (Figure 2) was fabricated using a commercially available 2μm Complementary Metal-Oxide-Silicon (CMOS) process [5]. A single heart cell is mounted between two silicon dioxide clamps attached to microbeams. The strain gauge is part of an on-chip Wheatstone bridge. When the cell contracts, it activates a strain gauge located at the base of one of the microbeams. The signal from the bridge is amplified on chip and recorded using an off-chip data acquisition system.

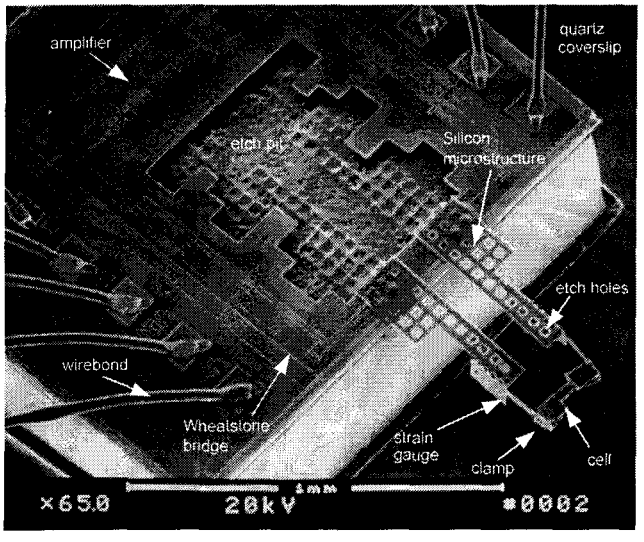

Figure 2. SEM photograph of MEMS heart cell force transducer.

The available layer stack for the CMOS process is shown in Figure 3. In this process Metal-Oxide-Semiconductor Field Effect Transistors (MOSFETs) are fabricated along with MEMS microstructures via photolithography and etching. The metals are used as electrical interconnects, while the polysilicon can be patterned into piezoresistive strain gauges and resistors. Since layers of polysilicon and metal are sandwiched between the oxide, the polysilicon and metal can be patterned such that electrical components are fully encased in the oxide. Thus, the oxide becomes part of the encapsulation packaging.

To create a folded, three-dimensional microstructure, the metal layers were also patterned into freestanding, exposed microbeams which act as mechanical hinges as well as electrical interconnects [6]. To encapsulate these beams, silicone rubber sealant was applied via a needle probe. The probe is attached to a 3-axis micromanipulator and is also used to assemble the microstructure once it has been "released" from the substrate. Release is achieved by etching the silicon substrate from underneath the oxide after the chips come back from the foundry.

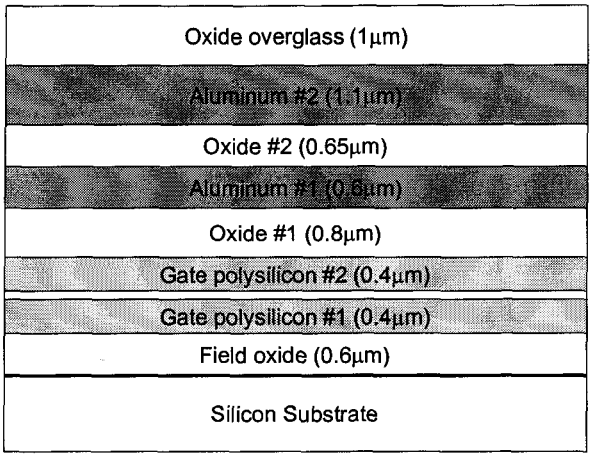

Figure 3. CMOS layer stack. 500Å of thermally grown SiO_2 is located between the polysilicon layers

"Vias" are holes patterned in the oxide layers by which layers of polysilicon and metal are connected to the substrate and/or each other. Etch windows are created by patterning successive vias on top of each other, thereby leaving the substrate exposed. The oxide microstructures can be released using any wet silicon etchant such as TMAH or KOH. However, due to the delicacy of most microstructures, a dry-phase etchant such as XeF_2 is preferred [7]. This isotropic gas-phase etchant offers high selectivity toward silicon dioxide and aluminum, and it eliminates any liquid meniscus forces or bubbles that can damage the microstructures. Etching is expedited by using etch holes (i.e. small etch windows patterned in a periodic array throughout the microstructure as shown in Figure 2).

Part of the device package includes interconnects to bring signals off the chip to a data acquisition system. Thus, before release etching the chip is bonded to a quartz coverslip on which ten 1μm-thick aluminum lines have been patterned. The signal pads on the chip are wirebonded to these lines, and insulated output wires are also attached. After release etching and device assembly, the wirebonds and the silicon substrate are encapsulated in epoxy (applied manually). Finally, the aluminum lines on the coverslip are encapsulated in enamel.

Although this encapsulation method is complicated and labor intensive, it adequately protected the device and allowed this device to function in a nutrient saline solution environment conducive to keeping the heart cells alive. Furthermore, it did not have any adverse effect on overall cell function. Using this device, heart cell forces in the micronewton range were measured [9].

RECONFIGURABLE SENSOR PACKAGING

As illustrated in the previous section, totally encapsulated MEMS devices have been used to make biological measurements. However, to expand the capability of MEMS biosensors, especially chemical

sensors, a more sophisticated, interactive package will be required. At NASA/JPL we are addressing this issue by creating a new reconfigurable biosensor package that will operate in liquid environments and allow sensors direct access to their environment. This type of packaging will allow MEMS biosensors to be an integral component for NASA's advanced life support systems and astrobiological studies that will require a host of environmental, chemical, and microbial direct sensing capability.

The reconfigurable package uses a protective polymer "door" that will open to the environment on command. As illustrated in Figure 4, a protective polymer covers the inlet until the underlying heater or actuator is activated. Upon activation the polymer will be moved either by mechanical, thermal, or chemical means. Different microactuation methods are currently being assessed such as shape memory alloys and magnetic microactuators. Inert polymers are being considered for the protection layer, such as teflon and viton.

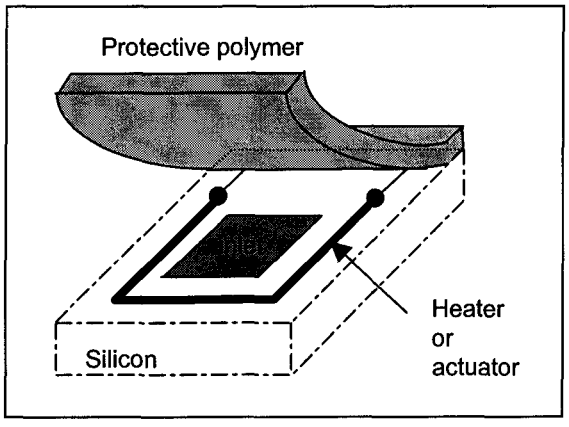

Figure 4. Schematic drawing of reconfigurable package under investigation at NASA/JPL. The inlet leads to a microfluidic circuit containing a sensor.

To date we have routinely spun thin films of Teflon AF (Dupont) onto silicon substrates. Typically the films are spun for 15 seconds at 2000 RPM. The films are roughly 0.8μm thick and are cured at 200°C for 2 hours. As expected, the films did not degrade when immersed in solvents. If not properly cured, we found that the films would delaminate when exposed to solvents. If the film delaminates, it comes off as a single sheet.

In addition to proper curing, we found that exposing the substrate to HMDS vapor prior to spinning on the teflon increased the teflon film adhesion. Films have been patterned using photoresist as a mask (AZ 5214 photoresist). Photoresist will not adhere to teflon unless the teflon surface is first roughened. Teflon surface roughening is achieved via an oxygen plasma at 200W for 5 minutes. The photoresist is spun on and patterned immediately afterwards, and the teflon is patterned using an argon plasma. 250W for 30 minutes was needed to pattern a 0.8μm-thick teflon film. A microchannel test pattern etched into teflon is shown in Figure 5. The photoresist was approximately 1μm thick. Etching times will vary due to nonuniformities in the film. Teflon film thickness can vary by as much as ± 0.2μm within a 100μm radius from a given point.

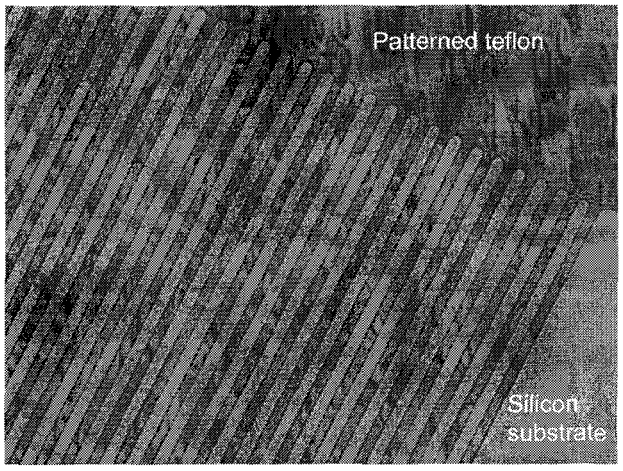

Figure 5. Optical microscope photograph of teflon microchannels patterned using photoresist. Channels are 35μm wide.

In a separate experiment we attempted to break teflon-coated membranes using metal heating wires. 300Å chromium and 2000Å gold were evaporated onto 1μm of thermally grown silicon dioxide. The Cr/Au was patterned into heating wires using a liftoff process and was subsequently covered with an unpatterned 0.8μm film of teflon. The metal heating wire is used to create a large thermal gradient across the silicon dioxide, initiating cracks. The tensile stress in the oxide is then utilized in creating out-of-plane motion, which tears the teflon. To form the membrane, the silicon substrate was etched from the backside using an SF_6 plasma in an STS deep-trench etcher. Membrane size varies from 200μm square to 1mm square. A representative test structure is shown in Figure 6.

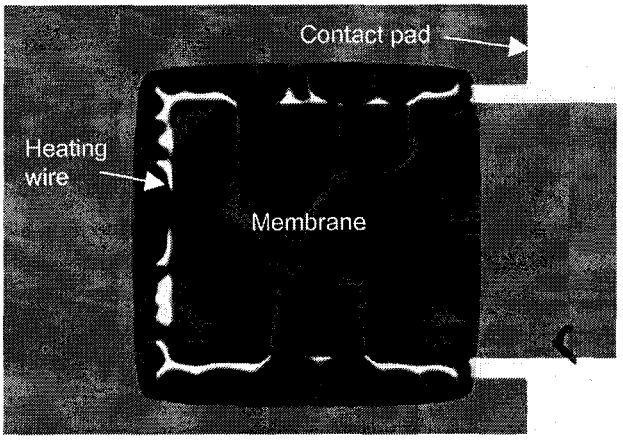

Figure 6. Optical microscope photo of teflon membrane test structure. Membrane shown is 500 μm x 500μm. The heating wire is 30μm wide.

Current was applied to the heating wire by touching two tungsten needle probes to each pad at a probe station. The test structure was incorporated into a circuit as shown in Figure 7. Preliminary results indicate that hundreds of milliamps are required to initiate oxide cracking. The tensile stress in the oxide film caused the attached teflon film to lift out of the plane of the wafer, but the amount of force was not high enough to tear the teflon. Higher tensile stress is needed and can be introduced either into the oxide or by adding a material layer with high tensile stress such as platinum or silicon nitride. Also, patterning the teflon into "doors" may facilitate lifting of the teflon during actuation. Both of these options are being pursued.

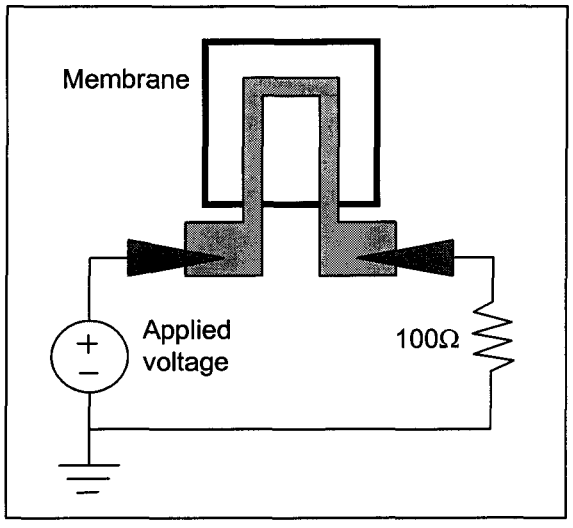

Figure 7. Schematic diagram of test circuit for membrane test structures.

Other methods of polymer actuation are also being considered, such as piezoelectric actuation. Electrochemical actuation may also be a viable alternative. Conjugated polymer actuators expand and contract when the polymer is electrochemically oxidized and reduced. They have proven to be robust actuators that function in ionic solutions (such as saline and other biological fluids) [9].

Electrochemistry may also be used to dissolve or burst the membrane. Electrochemical dissolution of thin metal membranes covering microreservoirs has been demonstrated [10]. The transformation of liquid into vapor can be achieved electrochemically as well. Sufficient vapor pressure will tear polymer membranes.

CONCLUSIONS

In this paper we describe several options for MEMS packaging for biomedical applications. Complete encapsulation, while more straightforward, may be not be suitable for devices that need access to the liquid environment to be assessed. To this end, we are pursuing a reconfigurable packaging scheme that consists of an inert protective polymer layer that can be moved or broken on command. To date, we have patterned thin films of Teflon AF using photoresist as a mask. Breaking or lifting a teflon-coated membrane has proven to be more challenging, but preliminary tests indicate that perhaps patterned teflon combined with more actuation force should yield favorable results.

ACKNOWLEDGMENTS

The research described in this paper was carried out at the Jet Propulsion Laboratory, California Institute of Technology, and the University of California, Los Angeles. This work was sponsored by the American Heart Association, Greater Los Angeles Affiliate, Grant-in-Aid #1059 GI-3, the National Institute of Health, #HL-47065, and the National Aeronautics and Space Administration Center for Space Microelectronics Technology under NASA UPN-632.

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